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(54) Wide aperture gradient set.

(57) An insertable coil (40) is inserted in a bore (12) of a magnetic resonance imaging apparatus. Primary field magnets (10) create a temporally constant magnetic field longitudinally through the insertable coil. A computer control (58) controls a radio frequency coil (44) and a gradient coil (42) to create magnetic resonance imaging sequences and process received magnetic resonance signals into image representations. The insertable gradient coil includes a central cylindrical portion (60) having a first circumference. A second portion (62) disposed toward a patient receiving end of the insertable coil has a second circumference which is larger than the first circumference. In this manner, the first, smaller circumferential portion is adapted to receive the patient's head and the larger circumferential portion is adapted to accommodate the patient's shoulders. For symmetry which eliminates magnetic field induced torques, a service end (68) matches and is symmetric to the patient end (62). The z-gradient coil windings (FIGURE 3), and x and y-gradient coil windings (FIGURES 4 or 5) are mounted such that a portion of the windings are on the first, smaller circumferential portion and a portion of the windings are on the larger patient and service end portions.

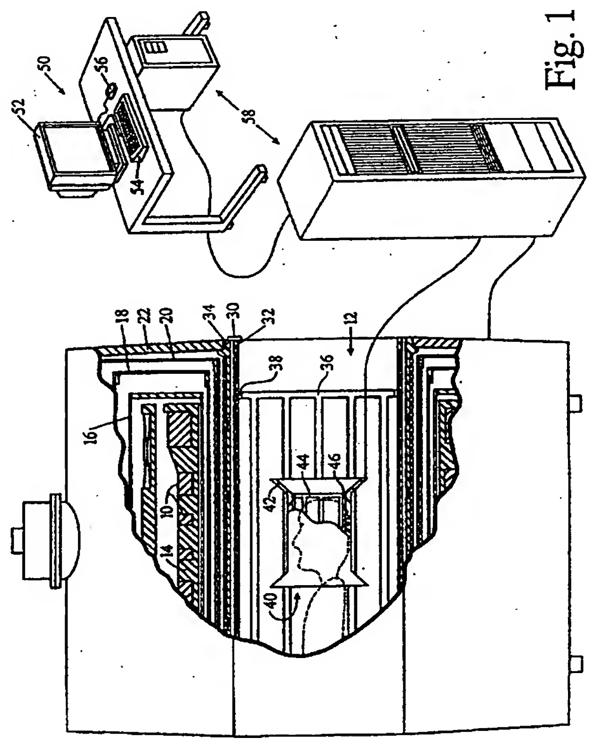

Fig. 1

EP 0 672 915 A1

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The present invention relates to the magnetic resonance imaging art. It finds particular application in conjunction with insertable gradient coils for high speed imaging techniques and will be described with particular reference thereto.

Magnetic resonance imagers commonly include a large diameter, whole body gradient coil which surrounds a patient receiving bore. Main field magnets, either superconducting or resistive, and radio frequency transmission/reception coils also surround the bore. Although the whole body gradient coils produce excellent linear magnetic field gradients, they have several drawbacks. With large diameter gradient coils, the slew rate is sufficiently slow that it is a limiting factor on the rate at which gradient magnetic fields can be induced and changed. Large diameter whole body gradient coils are too slow for some of the highest speed magnetic resonance imaging techniques. The energy stored in gradient coils is generally proportional to greater than the fifth power of the radius. Hence, large diameter, whole body coils require large amounts of energy. Further, superconducting main magnets have cold shields disposed around the bore. The larger the diameter of the gradient coil, the closer it is to the cold shields and hence the more apt it is to produce eddy currents. More shielding is needed to prevent the whole body gradient coils from inducing eddy currents in the cold shields than would be necessary for smaller diameter coils.

Due to these and other limitations in whole body coils, numerous insertable coils have been developed which are small enough to fit within the bore with the patient. Typically, the insertable coils are customized to a specific region of the body, such as a head coil. Traditionally, head coils have been a cylinder sized to accommodate the human head easily, e.g. 28 cm in diameter. Most brain examinations center around the portion of the brain that is substantially in the same plane as the eye sockets. In a symmetric coil, the magnetic isocenter is configured to be disposed in a common plane with the patient's eyes. For a symmetric coil, the magnetic isocenter is at the physical center of the coil.

As a general rule, the longer the cylindrical head coil, the larger the region over which the gradient is linear and the more linear the region is. However, the patient's shoulders are a limiting factor on the length of a symmetric gradient coil. The shoulders limit the isocenter to about 20 cm at the patient end. Thus, symmetric head coils have heretofore been limited to about 40 cm in length.

In order to achieve the beneficial effects of a longer head coil, head coils have been designed in which the magnetic isocenter is offset toward the patient from the physical geometric center of the coil. See, for example, U.S. Patent No. 5,278,504 of Patrick, et al. or U.S. Patent No. 5,177,442 of Roemer, et al. Although asymmetric head coils have beneficial effects on the linearity and the size of the linear region, the improvement is not without an offsetting difficulty. In the main magnetic field, the asymmetric gradient coil is subject to mechanical torques from the magnetic field interactions. In order to counteract these torques, the asymmetric head coils are mounted with rigid mechanical constraints. Even with substantial mechanical structures anchored to the main field magnet assembly, the torque still tends to cause at least mechanical vibration and noise. Adding additional windings to produce a counteracting torque commonly increases the electrical power requirements by about 50% or more.

Although conventional head gradient coils include a Maxwell pair for the z-axis or Golay saddle coils for the x or y-axes on the surface of a cylinder, others have proposed coils in which all windings do not lie on the cylinder surface. "Compact Magnet and Gradient System For Breast Imaging", S. Pissanetzky, et al., SMRM 12th Annual Meeting, p. 1304 (1993) illustrates a compact asymmetric cylinder coil bent up radially at a 90° angle at the field producing end of the coil. The coil is designed for breast imaging with the coil pressed up against the chest. "High-Order, Multi-Dimensional Design of Distributed Surface Gradient Coil", Oh, et al., SMRM 12th Annual Meeting, p. 310 (1993) attempts to optimize a gradient surface coil using current flows in three dimensions. One of the problems with the Oh surface gradient coil is that it was difficult to control linearity. Further, the coil was difficult to manufacture due to its complicated shape and high current densities.

The present invention provides a new and improved insertable gradient coil which overcomes the above-referenced problems and others.

In accordance with the present invention there is provided a gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the assembly comprising a central cylindrical region of a first circumference dimensioned to receive a region of interest of the subject to be imaged, the central cylindrical region carrying portions of gradient coils for creating linear magnetic field gradients through the region of interest, characterised in that the assembly comprises a first end region adjacent at least one end of the central region, the first end region having a second circumference which is larger than the first circumference, the gradient coils including windings on said first end region.

Thus, gradient magnetic field inducing windings adjacent the magnetic isocenter are of a first, smaller radius dimensioned to receive the head or other selected anatomical portion of the patient. Adjacent at least the patient end of the coil, the windings extend along a second, larger radius.

The gradient coil assembly may include at least two rings of bunched windings of the first, smaller radius.

The bunched windings may be connected with return paths which extend along the second, larger radius.

The gradient coils may include a plurality of thumbprint windings. The thumbprint windings may have the first, smaller radius adjacent the isocenter and may be flared to the second, larger radius adjacent at least the patient end of the coil.

The coil may be symmetric such that windings of the second, larger radius are disposed adjacent both the patient end of the coil and an opposite service end of the coil.

The coil may be of elliptical cross-section.

One advantage of the present invention is that it achieves better linearity by allowing an increase in length.

Another advantage of the present invention is that it provides better access for positioning the patient inside the coil.

Other advantages of the present invention include higher peak gradients and faster rise times.

Another advantage of the present invention is that it permits faster data acquisition and higher gradient frequencies.

Yet another advantage of the present invention is that it avoids excessive current density, particularly along patient and service ends of the coil.

Embodiments of the invention will now be described, by way of example only, with the aid of the drawings, of which:

Figure 1 is a diagrammatic illustration of a magnetic resonance imaging system including an insertable coil in accordance with the present invention;

Figure 2 is a diagrammatic illustration of a head coil in accordance with the present invention in relation to a human patient;

Figure 3 is a diagrammatic illustration of a z-gradient winding for the head coil of Figure 2;

Figure 4 is a diagrammatic illustration of one quadrant of an x or y-gradient fingerprint type winding for the gradient coil of Figure 3 laid out flat;

Figure 5 is a diagrammatic illustration of a bunched x or y-gradient winding for the coil of Figure 2;

Figure 6 is an alternate embodiment of the head coil of Figure 2;

Figure 7 is another alternate embodiment of the head coil of Figure 2, and

Figure 8 is yet another alternate embodiment.

With reference to Figure 1, a plurality of primary magnet coils **10** generate a temporally constant magnetic field along a longitudinal or z-axis of a central bore **12**. In a preferred superconducting embodiment, the primary magnet coils are supported by a former **14** and received in a toroidal helium vessel or can **16**. The vessel is filled with liquid helium to maintain the primary magnet coils at superconducting temperatures. The can is surrounded by a series of cold shields **18**, **20** which are supported in a vacuum dewar **22**.

A whole body gradient coil assembly **30** includes x, y, and z-coils mounted along the bore **12**. Preferably, the gradient coil assembly is a self-shielded gradient coil assembly that includes primary x, y, and z-coil assemblies potted in a dielectric former **32** and a secondary gradient coil assembly **34** that is supported on a bore defining cylinder of the vacuum dewar **22**. A whole body RF coil **36** is mounted inside the gradient coil assembly **30**. A whole body RF shield **38**, e.g. copper mesh, is mounted between RF coil **36** and the gradient coil assembly **34**.

An insertable gradient coil **40** is removably mounted in the center of the bore **12**. The insertable coil assembly includes an insertable gradient coil assembly **42** supported by a dielectric former. An insertable RF coil **44** is mounted inside the dielectric former. An RF shield **46** is mounted between the insertable RF and gradient coils.

An operator interface and control station **50** includes a human-readable display such as a video monitor **52** and an operator input means including a keyboard **54** and a mouse **56**. A computer control and reconstruction module **58** includes computer hardware and software for controlling the radio frequency coils **36** and **44** and the gradient coils **30** and **42** to implement any of a multiplicity of conventional magnetic resonance imaging sequences, including echo-planar imaging sequences. Echo-planar imaging sequences are characterized by short repetition rates and low flip angles. The processor **58** also includes a digital transmitter for providing RF excitation and resonance manipulation signals to the RF coil and a digital receiver for receiving and demodulating magnetic resonance signals. An array processor and associated software reconstruct the received magnetic resonance signals into an image representation which is stored in computer memory or on disk. A video processor selectively extracts portions of the stored reconstructed image representation and formats the data for display by the video monitor **52**.

With reference to FIGURE 2, the active gradient coil windings of the insertable gradient coil assembly **42** in the preferred embodiment are confined to a first cylindrical surface region **60** and an open patient end conical surface **62**. The cylindrical surface **60** has an isocenter **64** mid-way, a distance Z_{cp} from either edge of the cylindrical surface. The cylindrical surface has an interior dimension sized to receive the human head, preferably

with a radius ρ_c equal to about 15 cm. The RF screen and RF head coil reside inside of this diameter, between the subject's head and the gradient. The conical section **62** and a matching service end conical portion **68** flare from the radius ρ_c to a larger radius ρ_s.

This geometric shape for the gradient coil is symmetric, hence has an overall torque equal to zero. The imaging volume of the coil, i.e. the region with the best linearity and uniformity covers the entire human head and is centered on the brain. Due to the symmetry of the current density of the coil, its stored magnetic energy is less than the corresponding stored energy in an asymmetric gradient coil with the same specifications. The extended return path along the conical section permits lower turns densities.

To select the current winding pattern on a coil of this configuration, a sine Fourier-series expansion is performed. For the design of an axial or z-gradient coil, the current density is azimuthally oriented and varies only along the axial direction of the coil $J_\phi(z)$. As suggested by R. Turner, "Minimum Inductance Coils", J. Phys. E. Sci. Instrum. Vol 21, p. 948-952 (1988), expressing the z-component of the magnetic field B_z and the stored magnetic energy W in terms of the current density, the function E is defined as:

$$E(j_n^a) = W - \sum_{j=1}^{N} \lambda_j (B_z(\vec{r}_j) - B_{zSC}(\vec{r}_j)) \qquad (1),$$

where λ_j are the Lagrangian multipliers and B_{zSC}, represents the constraint values of the z-component of the magnetic field at N specified points. Minimizing E, a quadratic function of the current with respect to the current coefficients j_n^a, one obtains a matrix equation which $j_{n'}^a$ must satisfy:

$$\sum_{n'=1}^{\infty} j_{n'}^a \left\{ aL\pi \int_{-\infty}^{\infty} dk\ I_1(ka) K_1(ka) \psi_n(k) \psi_{n'}(k) \right\} =$$
$$-\sum_{j=1}^{N} \lambda_j \int_{-\infty}^{\infty} dk\ k \sin kz_j\ I_0(k\rho_j) K_1(ka) \psi_n(k) \qquad (2),$$

where the evaluation of the Lagrangian multipliers is done via the constraint equation. Inverting Equation (2), one can solve for $j_{n'}^a$, hence the current density. In these equations, a is the coil base radius and L is the coil length.

For manufacturability, the continuous current density is discretized. First, the continuous current distribution is divided into positive and negative current regions. Integrating the area of each region gives the amount of the total current contained in each region. Once the current for all regions of the cylinder is calculated, discrete current loops are placed in order to mimic the behavior of the continuous current pattern. Each region is then filled with discrete wires carrying a preselected amount of current which is the same for all discretized regions. The decision of the common amount of current for each loop is determined by considering the module of the current from all regions. In this case, the discrete coils can closely approximate the continuous current density. Specifically, each region of the continuous current density is divided into smaller segments which correspond to an equal amount of current. Then each wire is placed in the middle point of the segment in order to obtain an equal distribution from both sides of the segment when the magnetic field is calculated. In order to produce a current distribution which is confined to the cylindrical surface, a cut-off point in the axial direction is chosen at the distance Z_{cp} from the isocenter. With this point as the origin, the remaining end section of the cylindrical surface is tilted by an angle θ relative to the z-axis to form a conical surface. The relationship between the tilting angle θ and the axial, radial dimensions of the coil is:

$$\theta = \tan^{-1} \frac{\rho_s - \rho_c}{L/2 - z_{cp}} \qquad (3).$$

Because the rotation is a unitary transformation, no alterations to the coordinate system and its unit vector result. In this manner, a discrete current pattern for the axial z-coil which is confined to a 3D surface which is a combination of a cylindrical surface up to point z_{cp} and a conical surface for the rest of the coil length as illustrated in FIGURE 3 is provided.

With reference to FIGURE 4, the x or y-gradient coils start with the configuration of a traditional finite size x or y-gradient coil with radius ρ_c and total length L. The design of this type of gradient coil generates a gradient field which is odd-symmetric in the x or y-direction around the geometric center of the coil, respectively, while it is even-symmetric along the z and other of the x and y-directions, respectively. Due to the finite length of

the gradient coil, a Fourier-series expansion in terms of the sine and cosine series is performed. Due to the symmetric conditions along the z-direction, only cosine Fourier-series expansion is needed. Due to the pre-specified symmetric conditions, the current density lies on the surface of the cylinder and the resultant current density is constructed as a vector addition of two components - one along the axial direction $J_z(\phi,z)$ and the other along the azimuthal direction $J_\phi(\phi,z)$. Using the continuity equation in order to relate both components to the current density and expressing the z-component of the magnetic field B_z and the stored energy W in terms of either one of these two components of the current density, the function E is again constructed as follows:

$$E(j_n^a) = W - \sum_{j=1}^{N} \lambda_j (B_z(\vec{r}_j) - B_{zSC}(\vec{r}_j)) \qquad (4),$$

where λ_j are the Lagrangian multipliers and B_{zSC} represent the constraint values of the z-component of the magnetic field at N specified points. Minimizing E, a quadratic function of current, with respect to the current density coefficients j_n^a, one obtains a matrix equation which $j_{n'}^a$ must satisfy:

$$\sum_{n'=1}^{\infty} j_{n'}^a \left\{ \frac{aL\pi}{2} \int_{-\infty}^{\infty} dk\ I'_1(ka) K'_1(ka) \psi_n(k) \psi_{n'}(k) \right\} = \sum_{j=1}^{N} \lambda_j \cos(\phi_j) \int_{-\infty}^{\infty} dk\ k \cos kz_j\ I_1(k\rho_j) K'_1(ka) \qquad (5),$$

where the evaluation of the Lagrangian multipliers can be done via the constraint equation. Inverting Equation (5), a solution for j_n^a and hence the current density is obtained. Replacing this expression into the stored energy and magnetic field formulas, one acquires a final expression for the stored magnetic energy and the magnetic field in terms of the constraint points and the geometry of the system.

In order to obtain a discrete version of the above continuous current density, consider the continuity equation for the current density:

$$\nabla \cdot \vec{J} = 0 \qquad (6).$$

In analogy with the magnetic field where the vector potential is introduced, the current density can be expressed as a curl of the function $\vec{S}$, called a "stream function". Specifically:

$$\vec{J} = \nabla \times \vec{S} \qquad (7).$$

Because the current is restricted to flow on the surface of a cylinder of radius a and has only angular and axial dependence, the relation between the current density and the stream function in cylindrical coordinates is:

$$j_\phi^a(\phi,z)\hat{a}_\phi + j_z^a(\phi,z)\hat{a}_z = \frac{\partial S_\rho}{\partial z}\hat{a}_\phi - \frac{1}{a}\frac{\partial S_\rho}{\partial \phi}\hat{a}_z \qquad (8),$$

and S_ρ is found from:

$$S_\rho(\phi,z) = -a \int_{-\pi}^{\phi} d\phi'\ j_z^a(\phi',z) \qquad (9).$$

The contour plots of the current density are determined by:

$$S_\rho(\phi,z) = \left(n-\frac{1}{2}\right) S_{inc} + S_{min} \text{ for } n = 1, \ldots, N \qquad (10),$$

where N is the number of current contours, S_{min} is the minimum value of the current density, and s_{inc} represents the amount of the current between two contour lines. The determination of S_{inc} is:

$$S_{inc} = \frac{S_{max} - S_{min}}{N} \quad (11),$$

with S_{max} representing the maximum value of the current density. The contours which are generated by this method follow the flow of the current and the distance between them corresponds to a current equal to an amount of S_{inc} in amps. As illustrated in region z_{cp} of FIGURE 4, discrete wires are positioned in such a way as to coincide with these contour lines.

Looking now to the discrete current density portions which lie on the surface of the cone, a cut-off point along the axial direction is chosen at a distance z_{cp} away from the geometric center. With this point as the origin, the remaining part of the cylindrical surface is tipped by an angle θ relative to the z-axis. The relationship between the tilting angle θ and the axial, radial dimensions of the coil is:

$$\theta = \tan^{-1} \frac{\rho_s - \rho_c}{L/2 - z_{cp}} \quad (12).$$

Because the rotation is a unitary transform, no alterations in the coordinate system and the unit vectors are made. In this manner, a discrete coil pattern is generated for the transverse x or y-gradient coil which are defined in a 3D surface which is a combination of the cylindrical surface up to the point z_{cp} and a conical surface for the rest of the coil length.

To evaluate the magnetic field of the discrete current distribution, one uses the Biot-Savart law:

$$B_z = \frac{\mu_0 I}{4\pi} \int_{l_1}^{l_2} \frac{(\vec{dl} \times (\vec{r} - \vec{r}\,'))_z}{|\vec{r} - \vec{r}\,'|^3} \quad (13).$$

The evaluation of Equation (13) can be separated into two integration areas. The first integration area includes only the region of the cylindrical surface in which $\rho=\rho_c$. Thus, each current segment is only a function of the azimuthal direction ϕ and the axial direction z. The expression of the magnetic field which results from this current pattern is:

$$B_z = \frac{\mu_0 I}{4\pi} \int_{\phi_1}^{\phi_2} d\phi' \left\{ \frac{\rho_c^2 - \rho\rho_c\cos(\phi - \phi')}{[\rho_c^2 + \rho^2 - 2\rho\rho_c\cos(\phi - \phi') + (z \pm z')^2]^{\frac{3}{2}}} + \frac{\rho_c^2 - \rho\rho_c\cos(\phi + \phi')}{[\rho_c^2 + \rho^2 - 2\rho\rho_c\cos(\phi + \phi') + (z \pm z')^2]^{\frac{3}{2}}} - \frac{\rho_c^2 + \rho\rho_c\cos(\phi - \phi')}{[\rho_c^2 + \rho^2 + 2\rho\rho_c\cos(\phi - \phi') + (z \pm z')^2]^{\frac{3}{2}}} - \frac{\rho_c^2 + \rho\rho_c\cos(\phi + \phi')}{[\rho_c^2 + \rho^2 + 2\rho\rho_c\cos(\phi + \phi') + (z \pm z')^2]^{\frac{3}{2}}} \right\} \quad (14),$$

where

$$z' = z_1 + \frac{z_2 - z_1}{\phi_2 - \phi_1}(\phi' - \phi_1) \quad (15),$$

with z_1, ϕ_1, representing the coordinates of the origin for each line segment at the discrete current distribution, and z_2, ϕ_2 corresponding to the coordinates of the end point for the same line segment.

In the second region, each point of the discrete current pattern is a function of three variables (ρ, ϕ, z). Due to the cone, there is variation along the azimuthal direction ϕ, the axial direction z, and an additional dependence on the radial variation ρ. The expression of the z-component of the magnetic field for this portion of the equation becomes:

$$B_z = \frac{\mu_o I}{4\pi}\int_{\phi_1}^{\phi_2} d\phi' \left\{ \frac{\rho S_\rho \sin(\phi - \phi') + \rho'^2 - \rho\rho'\cos(\phi - \phi')}{[\rho'^2 + \rho^2 - 2\rho\rho'\cos(\phi - \phi') + (z \pm z')^2]^{\frac{3}{2}}} + \frac{-\rho S_\rho \sin(\phi + \phi') + \rho'^2 - \rho\rho'\cos(\phi + \phi')}{[\rho'^2 + \rho^2 - 2\rho\rho'\cos(\phi + \phi') + (z \pm z')^2]^{\frac{3}{2}}} - \frac{\rho S_\rho \sin(\phi - \phi') - \rho'^2 - \rho\rho'\cos(\phi - \phi')}{[\rho'^2 + \rho^2 + 2\rho\rho'\cos(\phi - \phi') + (z \pm z')^2]^{\frac{3}{2}}} - \frac{-\rho S_\rho \sin(\phi + \phi') - \rho'^2 - \rho\rho'\cos(\phi + \phi')}{[\rho'^2 + \rho^2 + 2\rho\rho'\cos(\phi + \phi') + (z \pm z')^2]^{\frac{3}{2}}} \right\} \quad (16),$$

with:

$$\rho' = \rho_1 + S_\rho(\phi' - \phi_1) \quad (17a),$$

$$z' = z_1 + \frac{z_2 - z_1}{\phi_2 - \phi_1}(\phi' - \phi_1) \quad (17b),$$

$$S_\rho = \frac{\rho_2 - \rho_1}{\phi_2 - \phi_1} \quad (17c),$$

where ρ_1, ϕ_1, z_1 are corresponding coordinates of the origin of the line segment at the conical region and ρ_2, ϕ_2, z_2 are corresponding coordinates of the end point for the same line segment. By employing the above steps, the discrete patterns of one quadrant of current distribution which is capable of generating 50 mT/m gradient strength over a 20 cm diameter spherical volume as shown in FIGURE 4 is created.

Similar results can be achieved using a bunched, rather than distributed, x or y-gradient coil. With reference to FIGURE 5, there are four, symmetric quadrants of bunched coiled windings which are symmetric relative to the isocenter **64**. Each bunched winding includes a plurality of generally semi-circular windings **70** adjacent a central plane of the cylinder **60** and a second, larger set of bunched windings **72** more distant from the central plane and again lying along the central cylinder **60**. Return windings **74** are mounted on either a conical or enlarged cylindrical region **76** of larger diameter than the cylinder **60**.

Various alternate embodiments are also contemplated. With reference to FIGURE 6, the coil may again have a central cylindrical portion **60** sized to fit the subject's head or other anatomical portion. A flared region **62** connects the central cylindrical region with outer cylindrical regions **66** of larger diameter. In a head coil embodiment, the larger cylindrical region **66** is of sufficient dimension to receive the patient's shoulders therein. For symmetry purposes, matching flared regions and cylindrical regions are preferably provided on both the patient and the service end **68** of the head coil. As another alternative, the central cylinder **60** can be elliptical to follow the generally elliptical cross-section of the human head and the outer larger cylindrical portion **66** is elliptical in a direction to match the aspect ratio of the human shoulders.

With reference to FIGURE 7, the flared portion **62** may be at 90° to the central cylinder **60**.

With reference to FIGURE 8, the insertable gradient coil need not have flared or enlarged portions at both ends. Rather, the central region **60** can connect a flared or enlarged region **62** at the patient end and an extended region **76** on the service end. The region **76** may be flared or tapered to a different radius than the patient end **62** or may be the same radius as the central cylindrical portion **60**.

Claims

1. A gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the assembly comprising a central cylindrical region (**60**) of a first circumference dimensioned to receive a region of interest of the subject to be imaged, the central cylindrical region (**60**) carrying portions of gradient coils for creating linear magnetic field gradients through the region of interest, characterised in that the assembly comprises a first end region (**62**) adjacent at least one end of the central region, the first end region (**62**) having a second circumference which is larger than the first circumference, the gradient coils including windings on said first end region.

2. A gradient magnetic field coil assembly as claimed in Claim 1, characterised in that the gradient windings extend circumferentially around the central region (60) and the first end region (62) with a plurality of turns for generating gradient magnetic fields axially along the coil.

3. A gradient magnetic field coil assembly as claimed in Claim 1, characterised in that the gradient coil includes at least one set of four gradient windings, at least two of the gradient windings being bunched coils which include arc segments (70) extending along the central region (60) and return paths (74) extending along the first end region (62).

4. A gradient magnetic field coil assembly as claimed in Claim 1, characterised in that the gradient coil includes at least one set of four gradient coil windings, at least two of the gradient coil windings being distributed coils each with its windings extending in generally a thumbprint pattern over approximately one quadrant of the central region and over approximately half of the first end region.

5. A gradient magnetic field coil assembly as claimed in any one of the preceding claims, characterised in that the central region (60) and first end region (62) are both circularly symmetric.

6. A gradient magnetic field coil assembly as claimed in any one of the preceding claims, characterised in that the first end region comprises a flared portion (62, 76) which angles outward from the central region (60) along an open, generally conical segment.

7. A gradient magnetic field coil assembly as claimed in any one of the preceding claims, characterised in that the first end region (60) includes a cylindrical portion (66).

8. A gradient magnetic field coil assembly as claimed in any one of the preceding claims, characterised in that the assembly comprises a second end region (68) of circumference greater than the first end region disposed adjacent an end of the central region opposite to the first end region, the gradient coils including windings on said second end region (68).

9. A magnetic resonance imaging system comprising: a primary magnetic field assembly (10) for generating a temporally constant magnetic field through a central bore (12) thereof; a whole-body gradient coil assembly (30) disposed around the central bore; a radio-frequency coil assembly (36) disposed around the central bore; a head coil (40) removably mounted in the central bore, the head coil comprising a radio-frequency coil winding (44); a magnetic resonance excitation and reconstruction means (58) for controlling the radio-frequency and insertable gradient coils for inducing magnetic resonance within the first cylindrical portion (60) and for receiving magnetic resonance signals therefrom and for reconstructing the received magnetic resonance signals into an image representation, characterised in that the head coil comprises a gradient magnetic field coil assembly as claimed in any one of the preceding claims

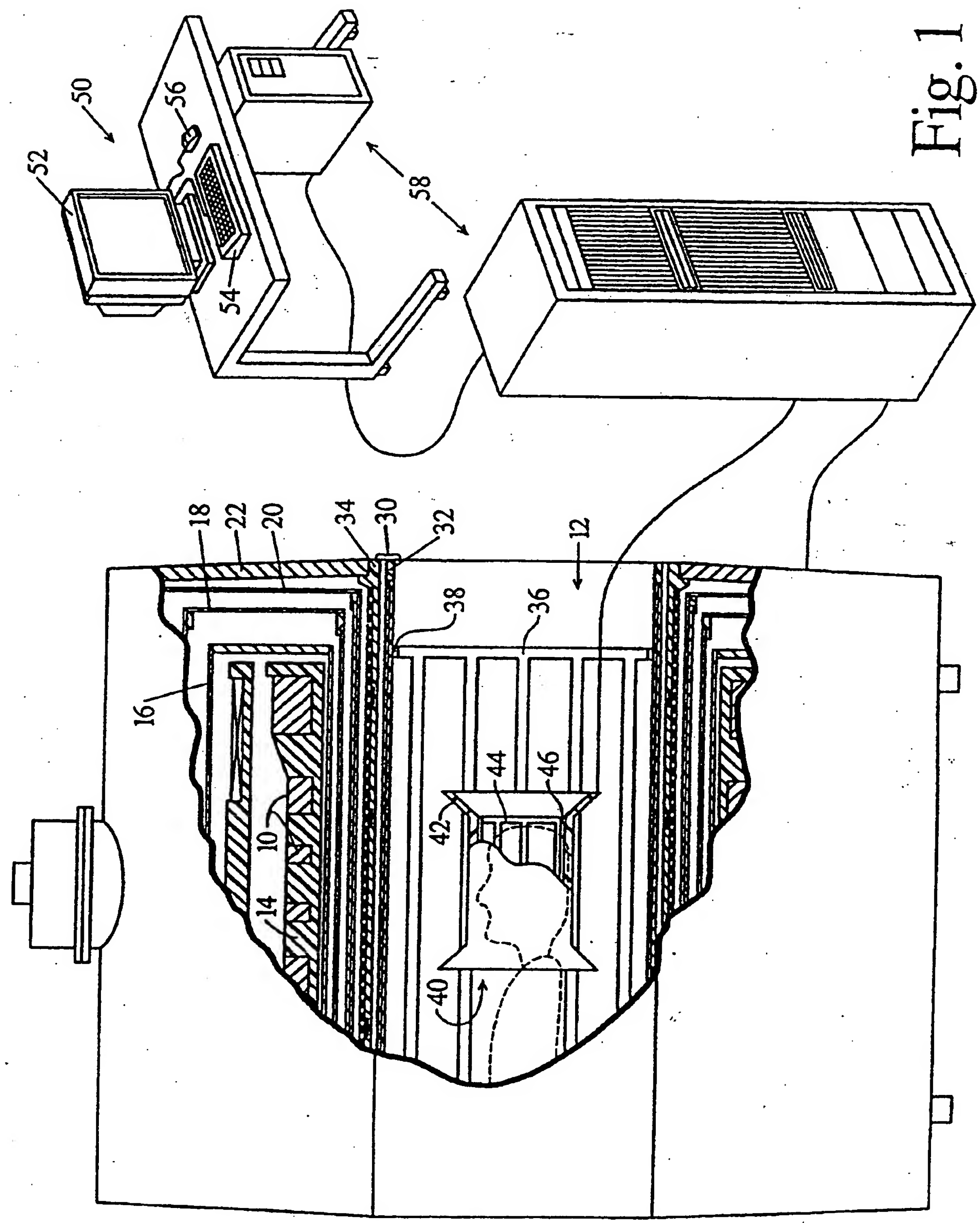

Fig. 1

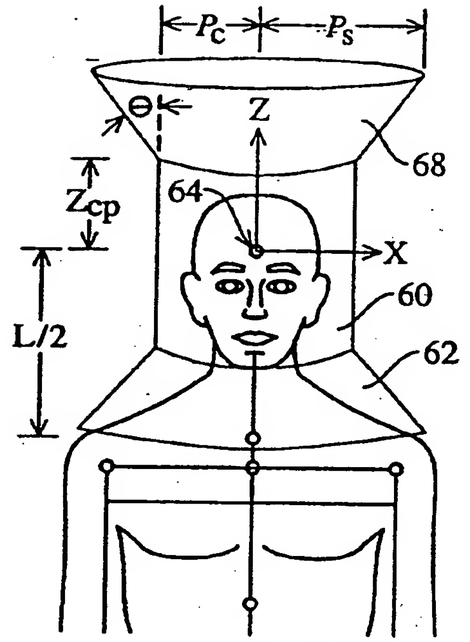
Pc
Ps
Θ
Z
68
Zcp
64
X
60
L/2
62

Fig. 2

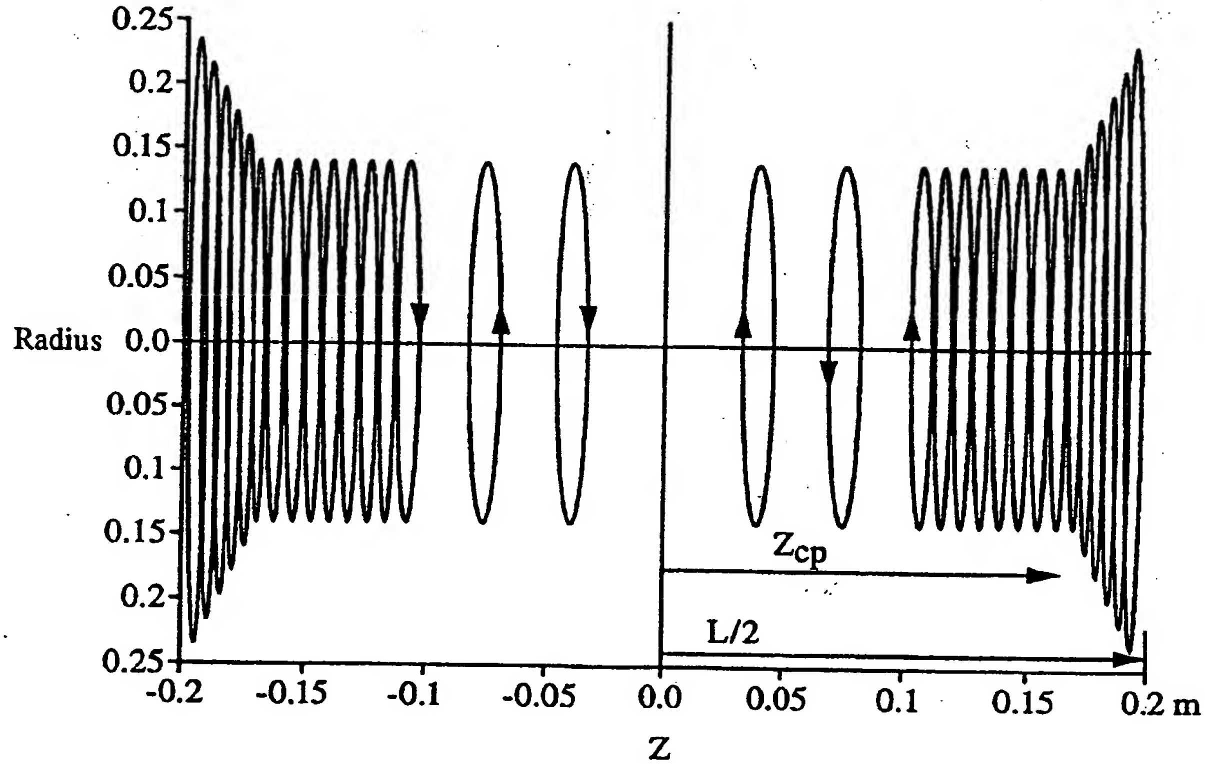
0.25
0.2
0.15
0.1
0.05
Radius 0.0
0.05
0.1
0.15
0.2
0.25
-0.2
-0.15
-0.1
-0.05
0.0
0.05
0.1
0.15
0.2 m
Z
Zcp
L/2

Fig. 3

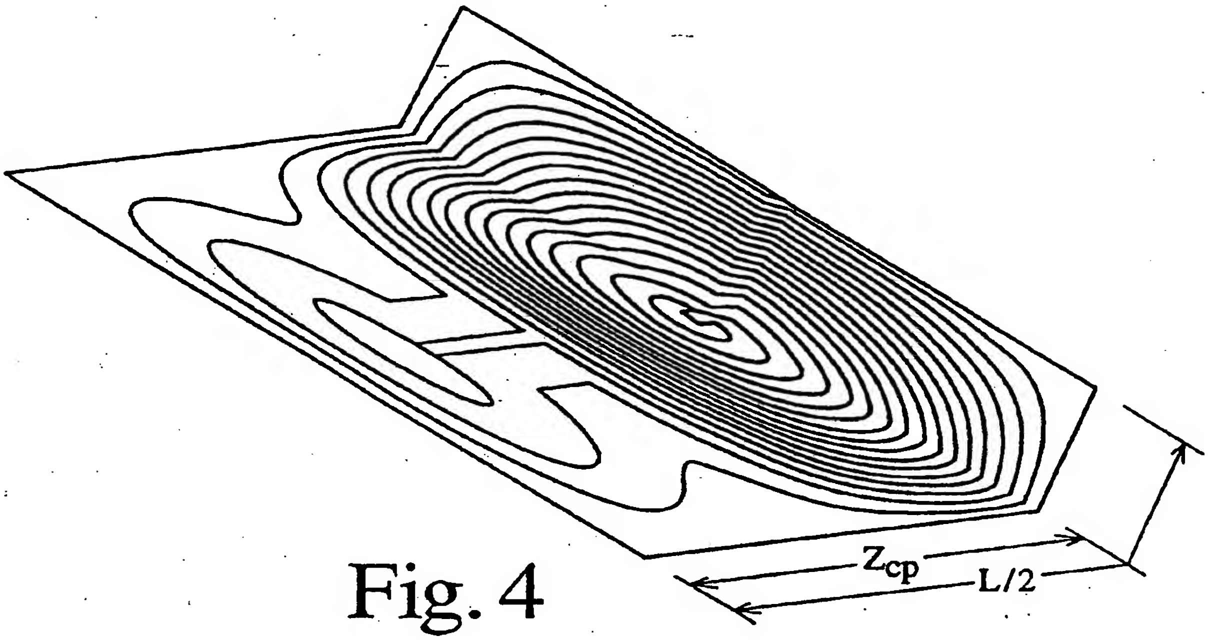

Fig. 4

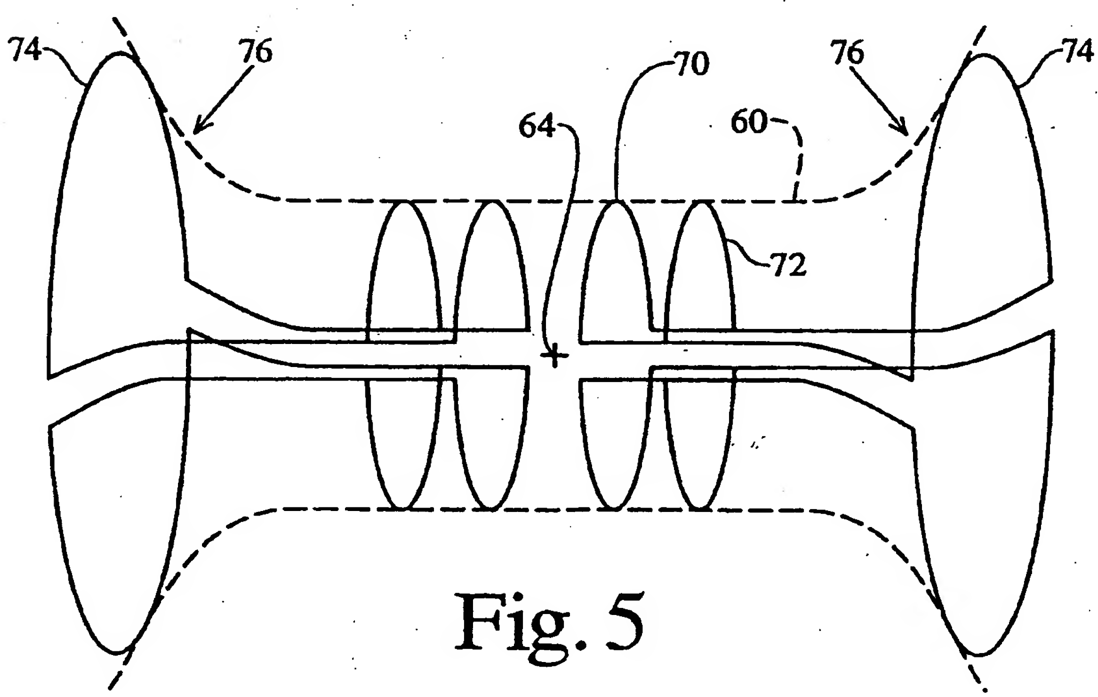

Fig. 5

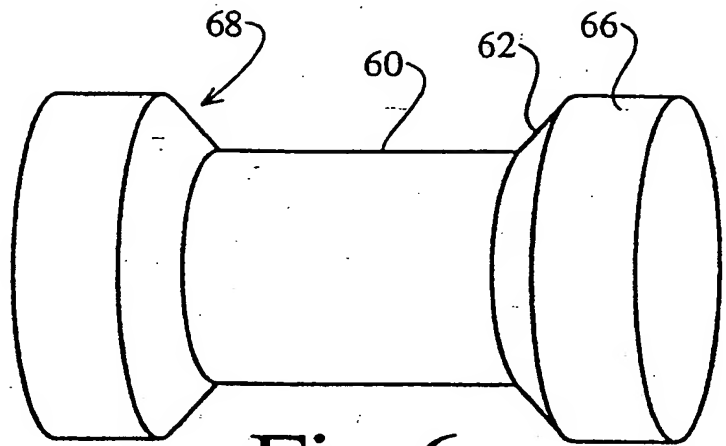

Fig. 6

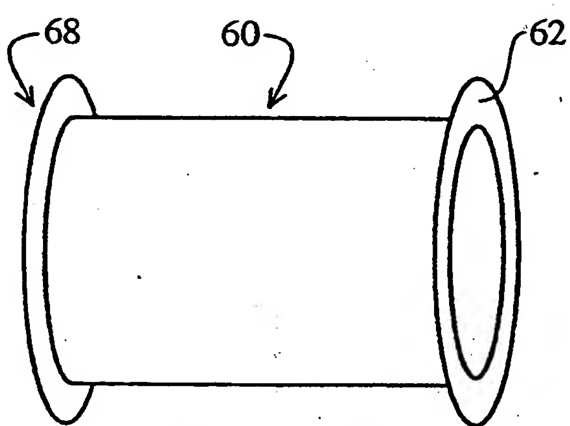

Fig. 7

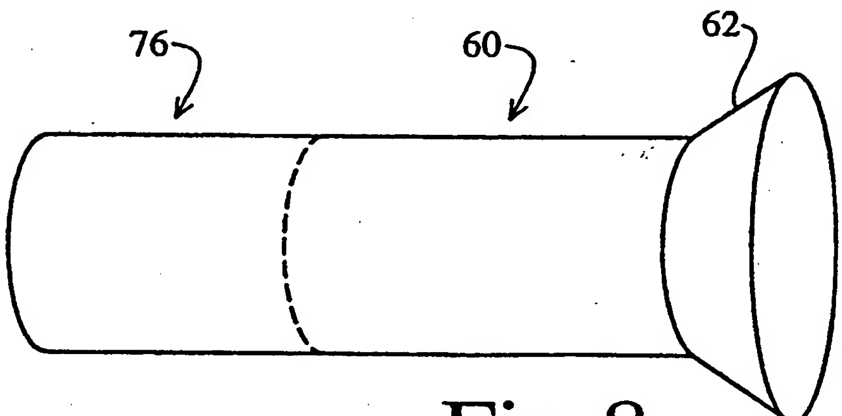

Fig. 8

European Patent Office

EUROPEAN SEARCH REPORT

Application Number

EP 95301163.2

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl. 6)
X	EP - A - 0 304 126 (PHILIPS) * Abstract; fig. 2,3 * --	1	G 01 R 33/385
D,A	US - A - 5 278 504 (PATRICK) * Fig. 2,4,5 * ----	1	
			TECHNICAL FIELDS SEARCHED (Int. Cl.6)
			G 01 R 33/00

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
VIENNA	31-05-1995	KUNZE

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
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